Optimal Conditions for Accelerated Imaging of Fractional Ventilation with Hyperpolarized Gas MRI
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INTRODUCTION: Pulmonary ventilation is an important marker in obstructive lung diseases, and its non-invasive imaging can provide useful information to investigate the severity of lung diseases and their response to intervention. Hyperpolarized gas MRI provides a noninvasive platform to directly image distribution of respiratory gas at a high resolution. However, quantitative imaging of ventilation still remains as one of the least developed areas using this imaging modality. An improved technique for fractional ventilation imaging was developed by authors [1] based on earlier work of Deninger, et al. [2], and was further adapted to large species (with pulmonary volumes comparable to humans) using accelerated imaging [3]. This work investigates optimal conditions for performing such measurements using parallel accelerated imaging.

METHODS: The multi-slice fractional ventilation imaging sequence is shown in Figure 1, for the case of three slices. Accelerated imaging was performed using parallel MRI and image reconstruction was done using the well-established GRAPPA methods [4] with a phased array receive coil. Fractional ventilation, ρ, defined as the ratio of the inspired volume to the total end-inspiratory volume, was measured on a regional basis using the technique described earlier [1], by fitting the signal buildup in breaths 1–N to a dynamic recursive model. For simulations, a 2D 64×64 image of the middle slice of a pig lung was used as the reference spin density map, M0(x,y) and M0(0)=0. The corresponding ρ(x,y) map, derived from the same pig study, was used as the a priori ρ map. Using α=5° and P0O2=140mbar throughout, a series of synthetic spin density maps were generated corresponding to a multi-slice ventilation imaging experiment using typical parameters in a pig study. Cartesian k-space acquisition scheme was then simulated according to the equation on the right; B(x,y); sensitivity profile for c-th coil, Nc; number of RF pulses. Various levels of complex random noise were added to the k-space signal. The spin density at each time point was sampled using a 1×4 phased array coil with identical sinusoidal sensitivity profile. The accelerated acquisitions were performed using ACL=8–32 and AR=2–4. Images from coils were reconstructed using GRAPPA algorithm and combined to obtain a single image for the corresponding time point, as shown in Figure 2. The effective acceleration factor of 64/N0c is calculated according to a given pair of ACL and AR. N0c=ACL+(64–ACL)/AR pulses. The resulting images were then fit to the fractional ventilation model to yield maps of ρ and ρ. Results were evaluated by: (i) RMS difference between the estimated and reference ρ maps, and (ii) correlation coefficient R for the voxel-by-voxel linear regression between the two maps.

RESULTS: ρ estimation error as a function of the applied α value was initially evaluated in a single imaged voxel used to determine the optimal flip angle over a range of N0c=24–64. The minimum error interval gets smaller for larger N0c as a result of the more pronounced effect of RF-induced depolarization. Optimal flip angle, αopt (α that minimizes Δρ) showed a highly linear behavior with respect to N0c (R=0.98–0.99 for r=0.2–0.4) and is confined to the range αopt=5°–7° for N0c=24–64. The variation of RMS error in the computed 64×64 ρ maps plotted was evaluated as a function of α, with the error normalized with respect to the minimum error (i.e. the optimal undersampling condition). Variation of normalized RMS error and R (linear regression between a priori and estimated ρ maps) both reach a local minimum and maximum, respectively, as a function of α (not shown for brevity). Based on the RMS error, ρ estimation error reaches a global minimum around N0c=31, beyond which the trend reverses, as shown in Figure 3, and increases with further undersampling. The nominal optimal scan parameters for this case are ACL=24 and AR=5, corresponding to an effective acceleration factor of ~2. A similar behavior is observed on the correlation coefficient of ρ maps. The technique was successfully implemented in pigs under mechanical ventilation as reported earlier [3]. A representative set of results is shown in Figure 4.

DISCUSSION AND CONCLUSION: In contrast to the single imaging voxel, it is evident that undersampling cannot indefinitely improve ρ accuracy, and there is a limit beyond which the information loss due to undersampling (e.g. reconstruction artifacts) outweighs the gain in reducing RF pulses and acquisition time. For assessment of accuracy of accelerated ventilation imaging, the normalized RMS error Δρ/ρΔρ2/ρ2max, not only reflects the effect of the number of RF pulses and noise, but also incorporates the inaccuracy introduced by undersampled image reconstruction artifacts. The minimum error condition for each (ACL,AR) pair also represents the corresponding optimal flip angle, αopt (not shown for brevity). It should be emphasized that this analysis only pertains to this representative case, and optimality conditions, in general, will be a function of other experimental details, including the number of parallel coils, imaging resolution, and achievable SNR.

Large species, humans included, breathe over a respiratory time scale of a few seconds (typical 4–8 sec breathing cycle at rest). Rodents have a respiratory rate of up to 10 times faster. The slower breathing rate of larger species means that certain signal decay mechanisms will longer be negligible in HP gas ventilation signal build-up (e.g. the oxygen depolarization effect (T1,1=18 s) induces a more prominent signal attenuation over the experimental time scale). Therefore in addition to diminishing the RF effect in ρ estimation, acceleration shortens the breath hold time necessary to acquire the images, thereby reducing the overall time and the associated O2-induced decay.